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Pre-operative planning for fracture fixation using locking plates: device configuration and other considerations

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Abstract

Most locked plating failures are due to inappropriate device configuration for the fracture pattern. Several studies cite screw positioning variables such as the number and spacing of screws as responsible for occurrences of locking plate breakage, screw loosening, and peri-prosthetic re-fracture. It is also widely accepted that inappropriate device stiffness can inhibit or delay healing. Careful preoperative planning is therefore critical if these failures are to be prevented.

This study examines several variables which need to be considered when optimising a locking plate fixation device for fracture treatment including: material selection; screw placement; the effect of the fracture pattern; and the bone-plate offset. We demonstrate that device selection is not straightforward as many of the variables influence one-another and an identically configured device can perform very differently depending upon the fracture pattern.

Finally, we summarise the influence of some of the key parameters and the influence this can have on the fracture healing environment and the stresses within the plate in a flowchart.

Introduction

Preoperative planning is critical to the success of any fracture fixation surgery. For any fixation device there are three key clinical requirements and consequent mechanical demands arising from them [1]: it must support fracture healing; it must not fail during the healing period; and it should not loosen or cause patient discomfort.

It is well recognised that an appropriate amount of interfragmentary motion (IFM) between fractured bone fragments is pivotal to healing; too much or too little can delay or inhibit fracture healing [2]. IFM is determined by the stiffness of the bone-fixator construct with stiffness defined as IFM produced on application of unit load. Moreover, to prevent failure, stresses within the implants should not be too high. Fatigue is normally a more likely cause of failure rather than a single traumatic event and the implant is more prone to fatigue failure if healing has been delayed [3]. Small increases in stress can, therefore, reduce significantly the number of cycles to failure of the fixation device [4]. High strains/stresses at the screw-bone interface are known to cause loosening around screw holes and entail a risk of infection [5, 6]. In addition, compromising the integrity of the bone due to screw holes or bone atrophy can lead to periprosthetic fracture during fixation or re-fracture after device removal [7].

In the context of metallic plates for fracture healing, preoperative planning must consider the different plate types available. The benefits of locking plates have been demonstrated clinically and experimentally [8, 9]. Several studies show that the use of locking screws can improve construct strength [10–12] and performance in osteoporotic bone compared to conventional screws [13–15]. On the other hand, studies have also shown that the pull-out strength of conventional screws increases with bone density [16, 17] which can result in equivalent or even better results than locked plating in healthy bone [14]. These differences arise due to two main factors: (1) The preloads involved in compression screw tightening increase strain levels at the screw–bone interface even before physiological loads are applied, whereas locking screws have negligible screw-tightening preload and resulting strains [13, 18]; and (2) During physiological loading, compression plating allows for frictional load transfer at the plate–bone interface; locked plating, on the other hand, transfers all physiological loads via the screw–bone interface [18]. In particular, the localised high tensile strains produced by conventional screws have been shown to be responsible for their poorer performance in osteoporotic bone [18]. Therefore, there should be a very clear distinction made between these two screw types; indeed, locking screws are not really screws in the conventional sense – they are more like bolts [19]. For example, the use of conventional screws can help reduce the fracture during surgery; on the other

hand, once a locking screw has been inserted, it prevents further distraction or reduction of the fracture [20].

It is well accepted that the majority of locked plating failures are due to inappropriate device configuration for the fracture pattern [21–25]. Several studies cite screw positioning variables such as the number and spacing of screws as responsible for cases of locking plate breakage, screw loosening, and periprosthetic re-fracture [3, 8, 26–29]. However, the significance of different variables and manner in which their variation affects mechanical behaviour of fixation constructs (and associated clinical expectations) is poorly understood; many of the findings in this respect are contradictory. To achieve the clinical requirements a well-planned device selection and configuration is essential, which in turn requires understanding the influence of different variables on the mechanical behaviour of bone-plate construct.

This aim of this study is to examine the role of different variables that influence pre-operative planning with a particular emphasis on device configuration, which is a key determinant for ensuring that the clinical requirements are met.

Philosophy of fixation

The decision whether to aim to for promotion of primary (direct) or secondary (indirect) bone healing needs to be made before any fixation device selection. Delayed fracture healing or non-union is very likely to occur when the fracture environment is not controlled to achieve one of these fixation philosophies [30]. Secondary bone fracture healing is the most common form of healing and the surgery required is less invasive and biologically damaging [2, 31]. To stimulate secondary bone healing, the initial post-operative interfragmentary movement (IFM) should be in the region of 10–40% of the total fracture gap [32, 33]. As the interfragmentary strain governs the healing process, the smaller the size of the fracture gap, the smaller the required movement. The appropriate value of IFM and resulting interfragmentary strain changes throughout the course of healing [34]. Primary bone healing is a much slower process requiring so-called ‘absolute stability’ of the fracture, and therefore aims to completely abolish the fracture gap; consequently, the required IFM tends to zero [34]. If any significant movement occurs in a small fracture gap, this results in very large strains and is disruptive to healing. Conversely, it is almost impossible to abolish relative movement between fracture fragments in a severely comminuted fracture pattern and therefore indirect bone healing should be sought [35]. Clinical studies demonstrate that using a lag screw to abolish movement in this situation conflicts with the goal of indirect healing leading to hardware failure [8]. One of the criticisms of

locking plates is that the final bone-plate construct can become overly stiff thereby delaying or preventing healing [20]. Therefore, the stiffness of the device should be carefully controlled.

Implant material selection

The interfragmentary movement (IFM) at the fracture site is largely governed by plate bending [36], and consequently plate stiffness needs to be carefully controlled to avoid it from being too high or too low and thereby detrimental to healing. Material choice is known to influence healing rates in distal femur locking plates, particularly in the period up to 12 weeks post-operative [37]. It is intuitive that titanium, with a lower Young's modulus than steel, produces greater interfragmentary movement (IFM). However, the increase in IFM produced by titanium compared to steel is not proportional to the difference in material stiffness as the plate is eccentric to the applied load [38]. The geometry of the plate, particularly the structural bending stiffness, also influences the IFM in a similar manner to the material stiffness.

Any implant will alter the natural load distribution within the host bone. Fixation devices are designed to redirect load and shield the bone from undesirable motion to allow the fracture to heal [39]. This redirection of load also results in other unwanted effects: for example, stress-shielding in some regions and strain concentration at the bone-implant interface. In locking plates, it has been shown that a more rigid screw material (e.g. steel with a higher Young's modulus than titanium) reduces the strain concentrations at the screw-bone interface [36] since they deform less in bending. The same applies to the stiffness of the plate itself – higher stiffness plates reduce screw-bone interface strain concentrations [36]. As material failure of bone and consequent loosening of screws is governed by strains, their concentration at the bone-screw interface needs to be limited. The concern of high interfacial strains with titanium in comparison to steel has been previously noted for unilateral fixators [40].

Device configuration: working length

One of the most important parameters regulating the device stiffness is the working length (also known as the bridging span), defined as the distance between the two innermost screws on either side of the fracture. Small working lengths in a simple reduced fracture can cause large plate stresses [4, 24, 41]; but in comminuted fracture patterns with a fracture gap, it is large working lengths that result in higher plate stresses [9]. This apparent contradiction has led to some confusion in the literature regarding the influence of working length. Bottlang et al. [42], for example, noted that the efficacy of working length, in terms of stiffness reduction, is “inconsistent and is gained at the cost of

construct strength". The mechanics for the two cases, one with a fracture gap and the other little or no fracture gap, can be explained as follows: When there is a fracture gap, the entire load is transmitted from one bone fragment to the other via the plate. In this case, upon load bearing, a higher working length results in flexible system leading to increased bending, higher plate stresses, higher interfacial bone strains and higher IFM. However, when there is no fracture gap the loads are shared between the bone and the plate. In this case a more flexible plating system (e.g. due to a larger working length) results in a lower load being transferred via the plate resulting in lower plate stresses and lower interfacial strains.

Unfortunately, the distinction between the performance of load-bearing and load-sharing locked plating systems is not fully understood. For example, some studies have attributed insufficient working length to higher plate stresses even for cases with a fracture gap [13, 19, 20]. This is explained [19] by applying identical angular deformation to the plate – a scenario in which a smaller working length will result in larger plate stresses.

A comparison between identical angular deformations with different working lengths is not clinically relevant, however, as plate stresses develop due to the identical loads (and not identical deformations) that the plate supports during weight-bearing. For a system with a large fracture gap, identical loads will cause larger plate stresses in a system with a larger working length (Figure 1a). This is due to the lower stiffness of the longer working length which results in larger deformations and plate strains. In some cases, when the fracture gap is small, interfragmentary contact can occur between fracture fragments [43]. If this happens, then the bone transmits load and the plate is shielded from stress increases as shown in Figure 1b [41, 43]. This is a load-sharing situation. In fracture patterns with wider gaps or comminution, interfragmentary contact cannot occur and the plate will have to transmit the full weight-bearing loads.

This means that an identically configured device can perform very differently depending upon the fracture pattern. For example, using three identical screw configurations, Stoffel et al. [41] found that larger working lengths produced the lowest plate stresses for small fracture gaps (1mm), but the highest plate stresses for larger fracture gaps (6mm). A flexible plate will deform more than a rigid plate under identical loading, developing larger strains in a load-bearing situation, but allowing load-sharing to occur and relieving strains if interfragmentary contact can occur.

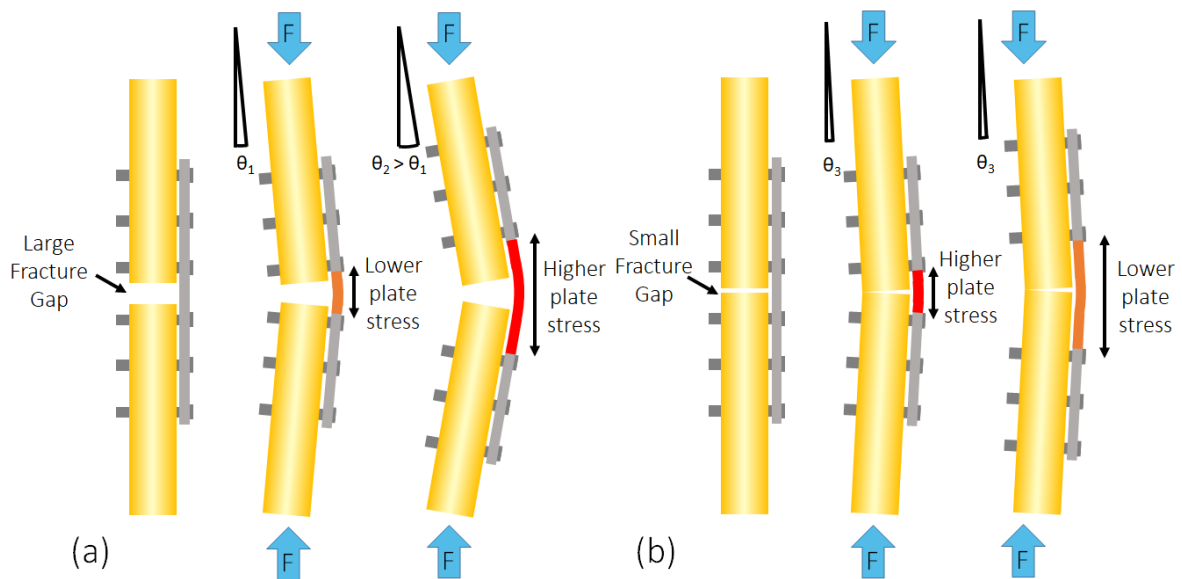


Figure 1 - A depiction of the influence of working length on plate strain in a bone plate system under applied axial load where: (a) there is a large fracture gap where no interfragmentary contact can occur (the plate is supporting all applied loads) resulting in larger plate stresses and larger angular deformation for system with larger working length; and (b) there is a small fracture gap which enables load sharing resulting in lower plate stresses with larger working length.

Device configuration: bone-plate offset

Locked plating is structurally similar to unilateral external fixation [1]; however, there is considerable difference in bone plate offset between locking plates and unilateral fixators. As the bone-implant distance is much smaller for locking plates compared to external fixators, the majority of the fracture gap motion is generated by plate bending [1] which also results in asymmetric fracture gap closure [44]. The authors have evaluated the influence of bone-plate off-set using the methodology from previous publications [36, 45]. The IFM at the near and far cortex for different values of offset are shown in Figure 2. Ahmad et al., [46] demonstrated that increasing the bone-plate offset can reduce the strength and stiffness of locking plate constructs. Techniques that utilise bending of the locking screws to increase the effective bone-plate distance have been described in a number of studies using: 'far-cortical locking screws' [47], 'dynamic locking screws' [48] and 'near cortical slots' [49]. These techniques have been shown to generate increased motion at the fracture site, however, they also produce larger strains at the screw-bone interface due to the larger lever arm [1, 50].

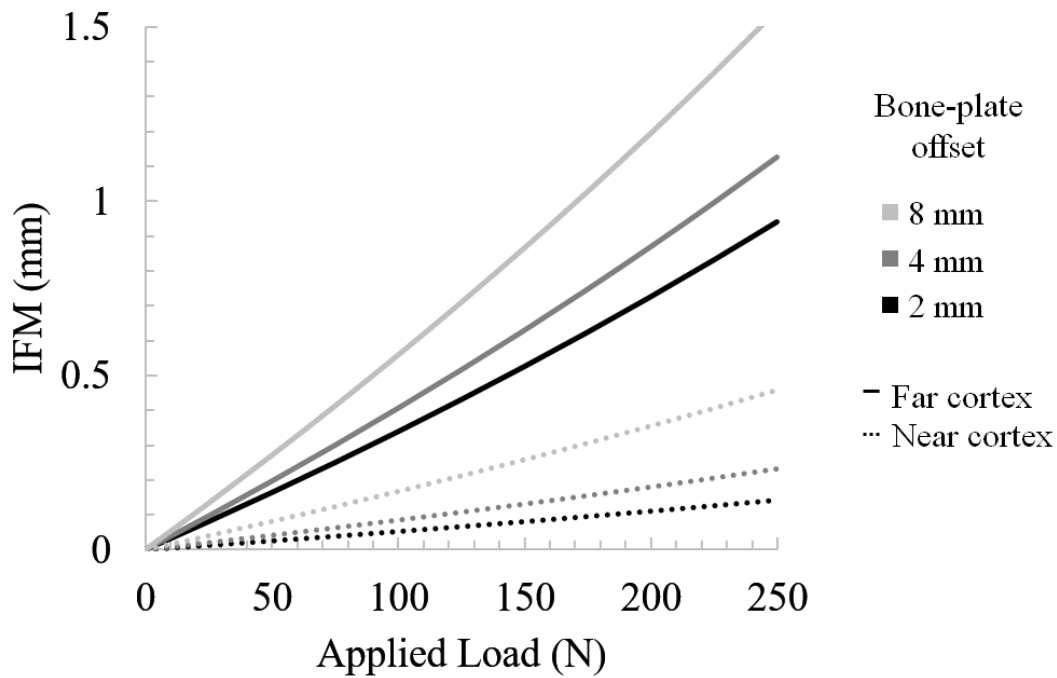


Figure 2 – The predicted interfragmentary motion (IFM) for a locked plate construct under axial loading depending upon the bone-plate offset [36, 45].

The influence of bone-plate offset in relation to the motion at the fracture site appears to have been overlooked by some studies in the literature. Two studies, Chao et al. [51] and Bottlang et al. [52], suggest that findings regarding working length are “inconsistent” citing articles by Field et al. [53] and Stoffel et al. [41] that evaluated the influence of screw position in plating (Figure 3). There are two key reasons why the study by Field et al. (1999) is not comparable to the study by Stoffel et al. (2003) and would, therefore, not provide consistent results. Firstly, the studies evaluated different loading regimes; Stoffel et al., used axial loading, whereas Field et al., considered bending loads. More importantly, Field et al. considered compression screws whereas Stoffel et al. considered locking screws. Conventional compression screws pull the plate towards the bone during fastening, thus eliminating a bone-plate offset. Similarly, in models where the plate was in contact with the bone, Chao et al., [51] and Hoffmeier et al., [54] both found that working length had very little influence on the fatigue life of constructs. Both studies acknowledged that the bone-plate contact may have masked any differences between screw configurations. If a plate is in contact with the bone, the effective working length is reduced and can be as small as the size of the fracture gap, and all screw configurations will produce very similar values of stiffness (Figure 3b). Therefore, this apparent conflict in the literature relating to the influence of working length on stiffness and fracture gap motion, can often be readily explained by the influence of bone-plate offset.

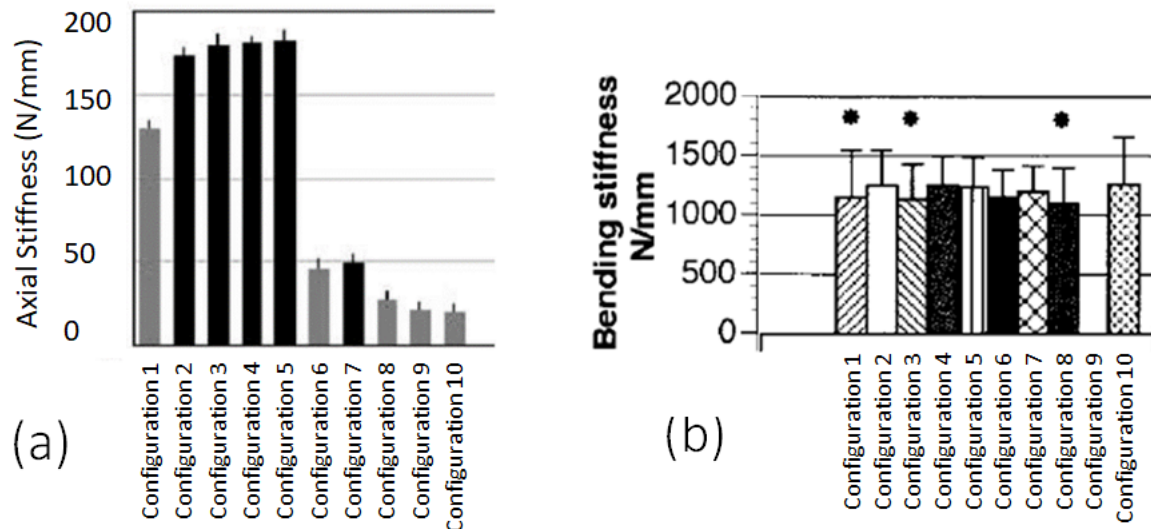


Figure 3 – (a) Axial stiffness values from Stoffel et al., 2003 who used locked plating with a bone-plate offset and (b) bending stiffness values from Field et al., 1999 who used compression plating with bone-plate contact. With permission from Elsevier. The screw configurations used in the two studies are not comparable and are simply used to illustrate the difference in the range of values found using different loading regimes.

Anatomical considerations, fracture pattern and boundary conditions

The anatomical location of the fracture can dramatically alter the demands placed upon the plate. In upper limb fractures plates are exposed to considerable torsional loads [55], whereas in the lower limb larger axial forces must also be resisted [56]. Lateral femoral plating causes the plate to be located a considerable distance from the mechanical axis due to the off-set of the femoral head. In a load-bearing situation, this can result in rapid plate breakage [34, 57]. It is important to note that a load-bearing environment can also occur in gap-opening situations (the opposite of the ‘tension-band’ principle), or where a displaced wedge fragment does not allow for load-sharing (Figure 4). Clinical studies, such as Strauss et al. [8], state that longer working lengths can “better distribute stresses” around the fracture site. This statement assumes that there is a load-sharing environment at the fracture site, in the case of a small fracture gap, as discussed earlier.

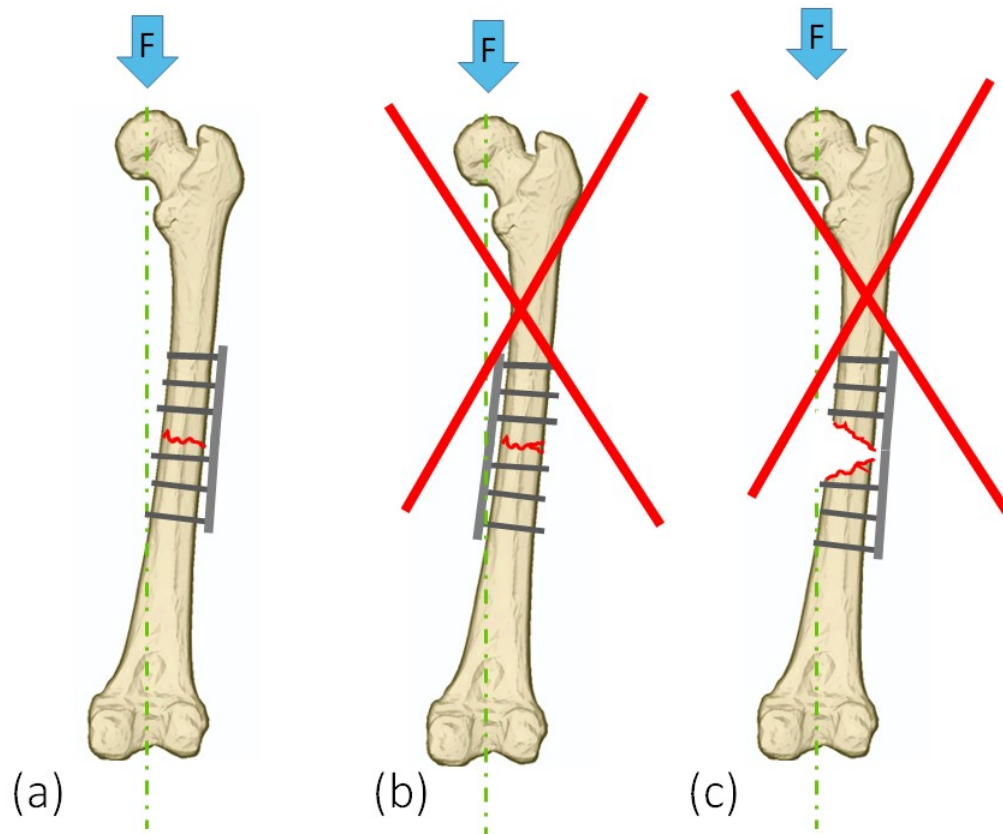


Figure 4 – Depiction of (a) the tension band load-sharing principle; (b) a gap-opening load-bearing situation; (c) a displaced wedge fragment load-bearing situation. The tension band situation (a) is safe for patient weight-bearing, however, weight-bearing in the other two situations (b and c) would result in very large stresses within the plate and a high risk of plate breakage.

The loading and boundary conditions can also influence whether bone-plate contact will occur. Chao et al., 2013 demonstrated that if the distal portion of the specimen is completely restrained while the proximal portion can rotate (Figure 5a), this results in significant shear motion which can lift the plate away from the bone if the working length permits (Figure 5b and c). In in vitro experiments methods of load application where the restraint condition is the same proximally and distally produce more symmetrical deformation patterns [38]. The deformation behaviour of plates under more physiological loading is still not well understood and it is not clear which in vitro experimental testing regime best predicts in vivo performance. The restraint provided by the knee, for example, is extremely difficult to recreate experimentally. Different experimental testing regimes such as bending, torsion or axial loading set-ups can yield dramatically different results in relation to the effects of device configuration. Moreover, even when using the same loading direction, the method of restraining the specimen can produce conflicting results [58].

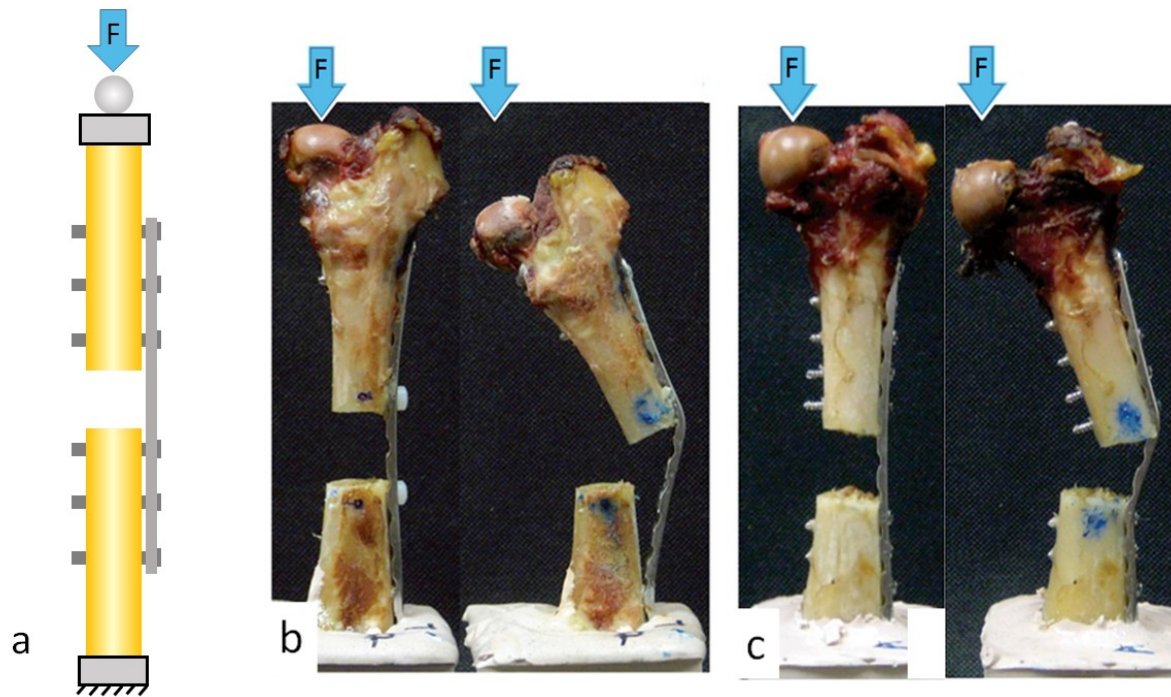


Figure 5 – (a) The experimental set-up adopted by Bottlang et al., 2009 and Chao et al., 2013. The results from Chao et al., 2013 show the influence of this specific loading scenario when using (b) a longer working length and (c) a shorter working length. Reprinted with from Biomed central Chao et al. BMC Veterinary Research 2013, 9:125 (open access)

Effects of fracture healing

A load-bearing system can become a load-sharing system with the progression of fracture healing. With increased callus formation, the bone will transmit an increasing proportion of the load [56]. If fracture healing does not occur or is delayed, then the plate must support all applied loading and plate breakage will eventually occur [22]. Previous experimental studies that have considered load-bearing scenarios have found that more rigid constructs perform better under cyclic loading [41, 51, 54, 59]. On the other hand, clinical studies show that more flexible plates promote faster callus formation; this includes the use of titanium instead of steel [37], but also the use of longer working lengths [60].

The authors have developed validated finite element simulations of locked plate constructs [61]. These models were used to evaluate the influence of callus formation on the plate stresses. Figure 6a shows the predicted IFM at the fracture site assuming an exponential increase in callus stiffness over time. Two screw arrangements are shown – a short working length resulting in a rigid plate-bone system; and a longer working length resulting in a more flexible plate-bone system. As expected, IFM reduces over the healing period. The stresses in both plates also decrease over the healing period, however, they decrease faster for the more flexible plate despite starting higher (Figure 6b). This

clearly illustrates the transition from load-bearing to load-sharing and the advantages of different working lengths for each situation. It can be inferred that, clinically, the reason longer working lengths reduce the risk of plate failure in the medium-to-long-term is because of increased load-sharing at the fracture site. Therefore, the presence of callus formation results in a similar influence of working length as the small fracture gap situation shown previously in Figure 2b (Figure 7)

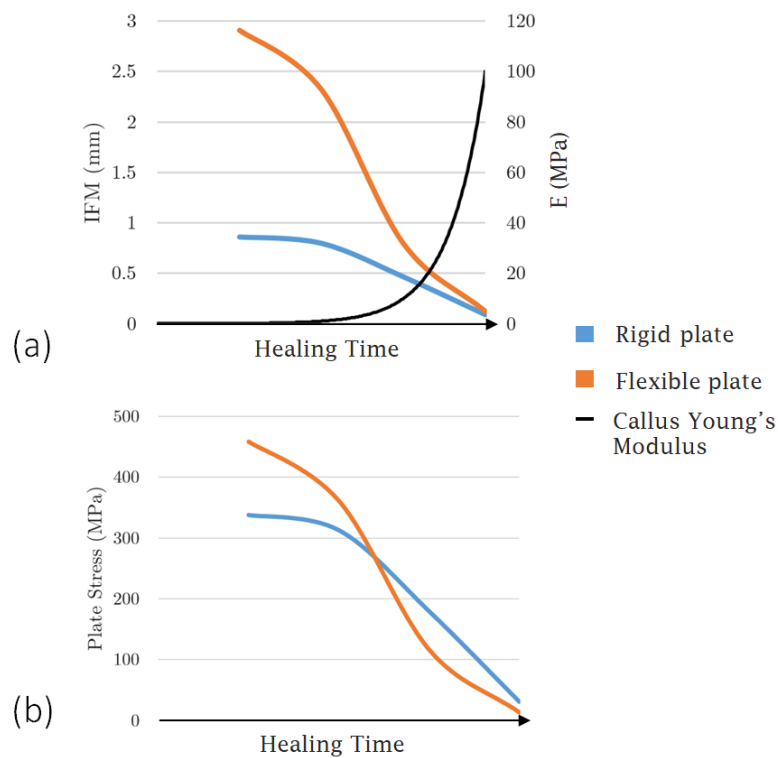


Figure 6 – The influence of rigid and flexible locking plates on (a) the interfragmentary movement (IFM) produced at the fracture site for a given callus Young's modulus; (b) the resulting plate stresses.

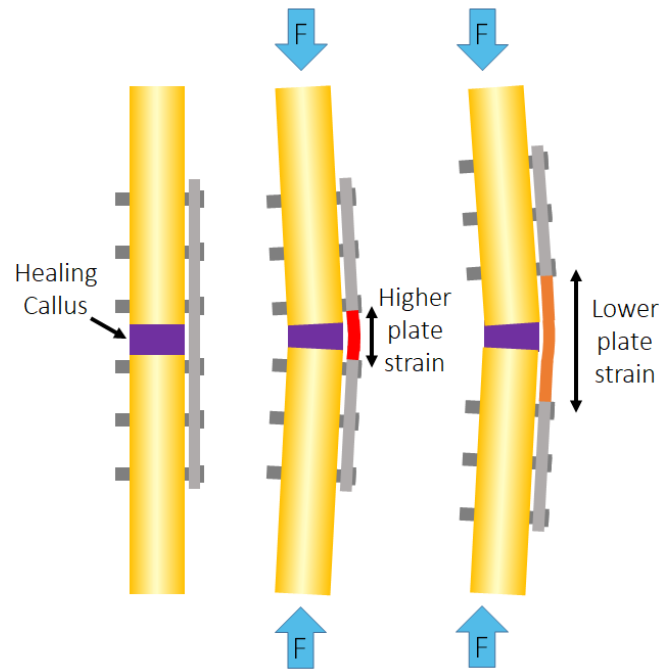


Figure 7 – the influence of callus formation at the fracture site on plate strain for different working lengths.

Bone quality

It has been clinically found that locking plates perform better than conventional plates in poor quality bone [13], whereas in healthy bone conventional plating can provide equivalent or even better results than locked plating [14]. The biomechanical reasons for this were described in a recent study [18]. Healthy bone has a thicker cortex and better mechanical properties compared to osteoporotic bone meaning that it can better withstand the transverse screw fastening loads. Interestingly, incomplete reduction or the presence of a fracture gap was found to increase the strains within the bone around the screws much more in conventional plating than in locked plating. Even in locked plating, complications resulting from screw loosening occur at least as often as plate breakage and delayed healing [24]. Strain concentrations within the bone around the screws can initiate loosening and result in infection and further loosening [40, 62–64]. Deterioration of bone quality, due to ageing or diseases, increase the risk of loosening [40, 65]. MacLeod et al. showed [36] that while bone quality affects loosening its influence on IFM is small. This has also been previously shown with respect to Ilizarov external fixators [66]. This implies that the required provision of appropriated IFM (or stiffness) can be considered independent from the requirement of loosening prevention. On the other hand, screw loosening is not independent of the stiffness of the bone-plate system; it has been found that the position of the screws on either side of the fracture affects these strain concentrations, with the working length being the most influential parameter [36]. In addition, the rigidity of the plate has also

been shown to be highly influential. Both of these parameters alter the stiffness of the fixation construct and therefore the environment at the screw-bone interface [1]. Generally, therefore, factors that increase the interfragmentary movement will also increase strains at the screw-bone interface. Strains, however, can also be manipulated by the positioning of the remaining screws (outside the working length). It should be noted that while for the majority of factors, bone quality does not affect the influence of screw placement, increasing the screw spacing on either side of the fracture can dramatically reduce strain concentrations in osteoporotic bone [36].

Optimising device configuration

Most plate failures are not due to a single traumatic event, but are the result of fatigue failure [4]; minimising stress in the plates and screws should, therefore, be one of the goals of internal fixation [4]. If healing progresses more rapidly, however, the demands upon the fixation are also reduced [67]. The most beneficial healing conditions require the correct level of fracture site movement – and this appears to be moderate axial motion and minimal shear movements [68, 69]. As discussed the axial motion produced by locked plating can be altered by many variables: working length, the offset distance between the bone and plate and the material properties of the plate [41, 46, 59]. Additionally, this review has demonstrated that there is confusion in the literature regarding the influence of some parameters in different fracture scenarios. This indicates that a more rigorous understanding of the mechanics of locked plating is required. Current screw positioning guidance includes the appropriate plate length and the number of screws to use (AO guidelines, [70]). Unfortunately, there are no guidelines regarding working length, which is known to be one of the primary determinants of interfragmentary motion in locking plates [41]. Better guidelines could inform the selection of an appropriate working length for a given fracture, which would reduce the likelihood of mechanical failure or delayed healing.

Considering the literature reviewed, the influence of working length, bone plate offset and fracture gap size, is summarised in Figure 8 [38]. As discussed, the apparent contradictory influence of working length is dependent upon whether load-sharing contact can occur at the fracture site under weight-bearing forces. The flow chart demonstrates that working length is only relevant when there is a bone-plate offset (left hand side of the flowchart in Figure 8). A fracture gap size of 1 mm was taken as a threshold where fracture gap closure might be expected under weight-bearing. This value is illustrative and will change for different device configurations.

If the plate is in contact with the bone (right hand side of the flowchart), then the effective working length is smaller than the distance between the screws closest to the fracture site. Additionally, in reduced fractures where primary bone healing is the goal, fracture reduction is more important than screw placement [29].

Therefore, if secondary fracture healing is the goal (or complete reduction is not achieved), screw placement can only be used as a method of controlling the healing environment or stresses in the plate if a bone-plate offset exists [4, 41]. It should be noted that, if fracture healing progresses the fracture callus will be supporting some of the load and the bone-plate construct will tend toward a load-sharing situation. If healing is delayed, however, the load-bearing situation would remain.

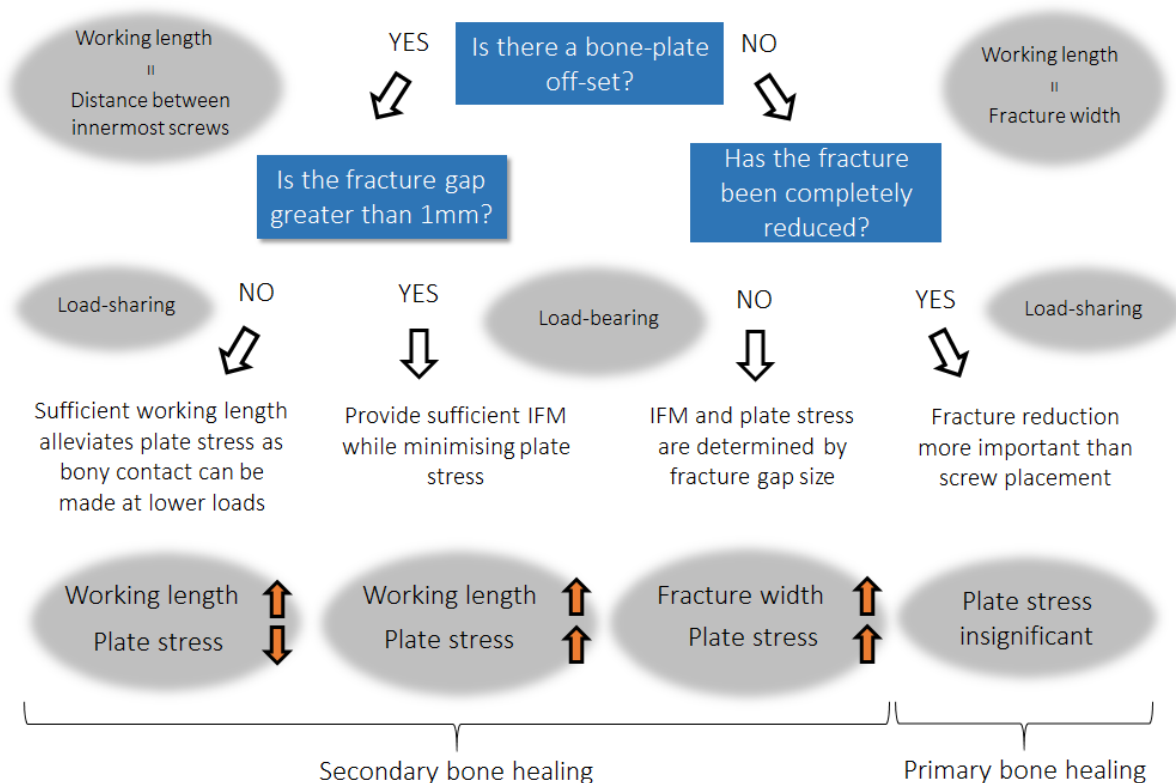


Figure 8 – Summary of the influence of different variables on plate stress including: bone-plate offset, fracture gap size and working length. Note: non-locking screws that compress the plate against the surface of the bone will always eliminate the bone-plate offset and, therefore, screw placement will have no significant influence on the plate stress under weight-bearing loads.

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